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REVIEW ARTICLE

Virtual Interactive Musculoskeletal System (VIMS) in orthopaedic translational research

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Summary The ability to combine physiology and engineering analyses with computer sciences has opened the door to the possibility of creating the "Virtual Human." This paper presents a broad foundation for a full-featured biomechanical simulator for the human musculoskeletal system. This simulation technology unites the expertise in engineering sciences and graphic modelling to investigate joint and connective tissue mechanics at the structural level and to visualize the results in both static and animated dynamic forms. Adaptable anatomical models including prosthetic implants and fracture fixation devices and a robust computational infrastructure for static, kinematic, kinetic, and stress analyses under varying boundary and loading conditions are incorporated on a platform, the Virtual Interactive Musculoskeletal System (VIMS), ideal for a cloud computing environment. A deployable database containing long bone dimensions, connective tissue material properties, and a library of skeletal joint system functional activities and loading conditions are also available that can be modified, updated, and expanded. An application software is available that allows end users to perform biomechanical analyses interactively. An example using the forearm and hand bone models plus a unilateral external fixator to study the distal radius fracture reduction in a virtual laboratory environment is highlighted to demonstrate this unique simulation technology in the field of orthopaedics.

Introduction

The concept of the "Virtual Human" aims at the understanding of human anatomy and physiology through simulation based on lifelike and anatomically accurate models and data. On a grand scale, the Virtual Human will lead to an integrated system of human organ structures that will explain various functional behavioural and

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activities of a "reference human." In recent years, the explosion of science and technology, creating an overlap between biological sciences and engineering know-how, has made Virtual Human a reality. This paper introduces the development and applications of a modelling and computational software package for the human musculoskeletal joint system, which enables the execution of a wide spectrum of biomechanical analyses under simulated or experimental environment. In addition to the visual attraction, it is an integration of physiological models, computer graphics, and analysis tools to determine the effects of pathological, ergonomic, and environmental conditions on the human musculoskeletal joint system. This effort represents a transdisciplinary collaboration among bioengineers, computer scientists, and physicians with multiple applications including medical education, translational research, and patient care—a precursor to the grand challenge of the "Virtual Human" concept.

This innovative concept and work have long been overlooked in the field of biomedical research, but it now represents a major force among a growing number of investigators in the traditional biomechanics discipline with the added strength of new engineering technology. Engineers have been working on adapting and refining the "Virtual Reality" (VR) concept for model analysis and data presentation from two-dimensional (2D), 3D, and even 4D space through system simulation and graphic visualization. The well-known flight and vehicular simulators in the past and medical procedure emulator at present provide realistic environmental and human-factor conditions for skill training and to monitor physiological responses. However, the engineering aspects of VR differ from those used in the fields of skill training, games industry, entertainment, and advertising. In addition to visual, tactile, and sensory requirements, bioengineering models must also be accurate, quantitative, computational, and interactive. These fundamental premises served as the underlying prerequisites of the present development and application. The recent advancements in surgical navigation and robotic systems are the direct outcome of this vision.

The Virtual Interactive Musculoskeletal System (VIMS) is a highly versatile simulation tool that provides information in an attractive, user-friendly, and easy-to-understand graphic environment while allowing computational algorithms embedded in the software architecture. Although these models and tools require enormous computer memory, processing speed, and graphic power, these are ideal and natural for the emerging technology of cloud computing (CC). This musculoskeletal biomechanics simulation system was built on proprietary software VisModel and VisLab (original products of the Engineering Animation Inc., Ames, IA, USA, now the property of EDS, Houston, TX, USA). It is divided into three integrated components—the "VIMS-Model," the "VIMS-Tool," and the "VIMS-Lab"—where each can function independently for a specific application, while also permitting collaborations through "VIMS.Net" (Fig. 1). In order to handle model variation among the normal population and pathologic effects, homogenous, multidimensional, and nonparametric scaling techniques are required based on the generic model forms. With the CC technology and virtually unlimited storage capacity, individual models based on the patient's imaging data can now be processed offline for treatment selection and planning service. These advancements were motivated by the limitations and demands of biomechanical analyses of musculoskeletal systems [1—5].

Multibody dynamic analysis of the musculoskeletal system has not received the attention it deserves partly because of the modelling and analysis difficulties involved. Furthermore, the internal muscle, ligament, and joint forces responsible for producing limb segment external loading and motion are still largely unknown. The redundancy of the control variables in the anatomical system and the distribution of the limb/joint forces among the tendons, ligaments, and articulating surfaces were only approximated using an optimization technique without adequate validation [5—8]. Incorporation of graphics with the model and results visualization offers definite advantages, but such advances have only been attempted and rectified recently. Although this proved to be a useful tool in modelling the system and in interpretation of the results, no comprehensive and in-depth interactive graphics capabilities were available to execute the analyses when the skeletal system is interfaced with implants or fixation devices. Buford et al. [9] used interactive 3D line drawings in a kinematic model of the hand. Later, a more attractive 3D surface model was introduced to calculate muscle-tendon paths in a biomechanical simulation environment [10]. Interactive graphical simulation software for modelling of the lower extremity was developed but mainly for static situations [11,12]. The models presented in this paper utilized rendered and shaded 3D graphics for display and allows the user to interactively set muscle paths and joint angles through a graphical interface. Although there are several commercially available modelling software in the market, none appear to provide the interactive capability particularly suited for biomechanical analyses.

A user-oriented network, the "VIMS-Net," subscribing to the same CC environment where VIMS will be managed and serviced can be established on the Internet to encourage close collaborations among different investigators in the musculoskeletal biomechanics community. This integrated software system and model database can impact on the learning of functional anatomy, the creation of virtual laboratories for biomechanical analyses without the use of animals or cadaver specimens, the development of patient-specific and device-based models for preinterventional planning in bone fracture management, limb lengthening, skeletal deformity correction through osteotomy, joint replacement, and radio frequency tumour ablation in orthopaedic oncology. Simulation-based teaching and skill training using virtual instruments and environment, and the establishment of a visual feedback and biomechanics-based system for computer-aided surgery and computer-aided rehabilitation—now popularized under the catchy terms of "surgical navigation" and "robotic surgery," etc.—can be established. Broader applications of this technology have already reached the advertising, litigation, financial, architectural, and entertainment fields.

Graphic-based model development—"VIMS-Model"

In essence, graphic-based models through simulation can bring the anatomical data "alive" and, through

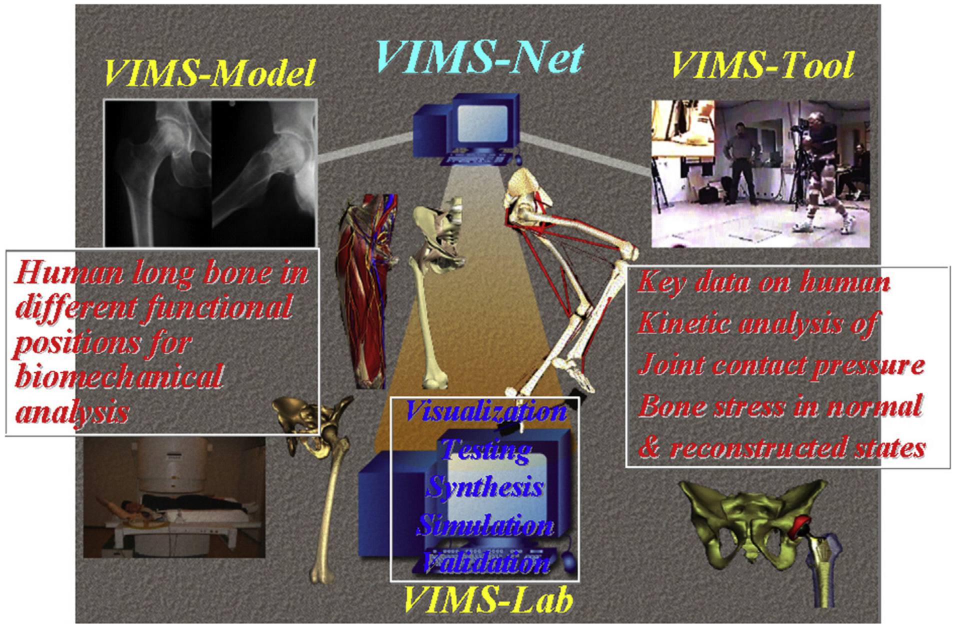

Figure 1 The structural flowchart and software platform of the Virtual Interactive Musculoskeletal System (VIMS) and database for biomechanical analyses.

biomechanical analyses, assess how the limb segments meet the functional demands of static and dynamic activities. Initially, anatomic data of the musculoskeletal system must be acquired and assembled into a model suitable for analysis and results visualization. Anatomic parameters related to joint function are quantified, including bone and soft tissue volumes, masses, and their relative orientation to one another. However, to accommodate soft tissue deformation in a model during activity is currently unavailable and should be developed in the future. The available database in VIMS-Model includes generic anatomic, orthopaedic implant, and device surface models necessary for biomechanical analysis. These models and database can be stored in a CC environment and retrieved by the users for different applications.

Geometric and material data acquisition

The "Visible Human" [14] is a set of volumetric image data of the human anatomy from two cadavers serving as the main source of the generic models stored on VIMS-Model library. The boundary-seeking algorithm provided by the commercial software, VisModel, was used to map out the profile of the 3D anatomic components in order to volumetrically reconstruct their surface shape. Computed tomography (CT) data were retrieved and analysed to build the voxels layer by layer according to preset grey level threshold to reconstruct the solid model for long bones containing different material properties and geometric irregularities. A database on isolated long bones from different populations combined with structural and material properties was used for comparative studies [15]. Other model databases available in the literature and from published reports can be incorporated and stored in the cloud. These data combined with the available scaling algorithms and personal imaging dataset provide the capability of creating individual patient-based models for treatment selection and intervention planning.

In soft tissues, the cross sections of these anatomic structures are outlined along their lengths, so that the centroidal lines of these tissue structures can be traced in 3D to define their line of action for biomechanical analyses. The muscle's physiological cross-sectional area [5] is included as an important parameter to determine muscle stress during static and dynamic activities. Muscle length and volume data are combined with their density values reported in the literature to estimate masses and moments of inertia for limb dynamic analysis. For cartilage, menisci, labrums, rotator cuff, and capsules, the detailed Virtual Human datasets are used to quantify their geometry in the models mainly for computational purposes. The articular cartilage thickness is an important parameter required in the intra-articular contact stress calculation. For the other soft tissue components, their fibre bundle orientation and insertion site are important for joint loading analysis. Although these soft tissue parameters are needed for biomechanical analyses, the lack of data on their shape and orientation changes in different joint positions in static and dynamic activities makes the geometric models unreliable in muscle force prediction. In addition, this deficiency prevents a graphically realistic presentation of the full limb in action during dynamic activities. Further development is thus needed in this respect.

Models for orthopaedic applications

In addition to musculoskeletal models, the VIMS system library also contains joint replacement implant models and bone fracture fixation devices for kinematic analysis and stress/strain analyses to aid users who wish to study their design, surgical placement, and clinical performance through simulation studies. Several generic models available within the VIMS-Model library are described here to illustrate their utility.

Full skeleton model

A full human skeleton model was adapted from a commercial source and modified by Engineering Animation Inc. as a general-purpose surface model (Fig. 2A—C). Local coordinate systems are imbedded in each skeletal component, which can be manipulated for animation purpose under given motion data. The surface shape represented by small polygons is fixed to the local coordinate system to facilitate rigid body motion analysis and animation. Through the use of surface smoothing, colour, shadow, and texture manipulation software, this interconnected model served the purpose of vivid visualization of the skeletal system in response to impact or under prescribed or recorded motion. This simplified model contains several movable joints with estimated degrees of freedom. No relative motion is permitted within the spine, trunk, hand, wrist, mid, and hind foot. However, these are available in localized joint models storage and can be incorporated when necessary. This interpolated skeletal model was used to animate human movements in normal functional activities and sports actions based on measured or calculated kinematic data for visualization purposes [16]. An example of the animated data related to a hip replacement patient is available in Video S3 in the Supplementary material online.

Shoulder musculoskeletal model

Detailed musculoskeletal models for the shoulder were constructed from cadaver specimens using their CT (for skeleton components) and magnetic resonance imaging (for muscles) data [17]. For other soft tissue details, the cryosectioned images were also used. These are surface models although they provide the layered muscular, neurovascular (the brachial plexus), and all underlying skeletal structures in a composite assembly, which can be visible in 3D in a sequential and animated form (Fig. 3A). These models were used for several kinematic and functional anatomy studies (Fig. 3B—D), and they provided the basis for muscle joint force analysis and joint contact stress and ligament tension in various activities (Fig. 3E) [18]. An animation on the layered anatomy of the shoulder with emphasis on the brachial plexus is available in Video S2 in the Supplementary material online).

Musculoskeletal model of the pelvis and hip

A composite surface model of the pelvis and all muscles across the hip joint was developed using the whole body database generated at the Johns Hopkins University, Biomechanics Laboratory and from the Visible Human Dataset available on the Internet (Fig. 4). In addition to illustrating the gross anatomy of the pelvis and the femur, this model was used to study hip joint contact stress during activities of daily living based on the living patients' data telemetered from their instrumented hip prosthesis [19,20] (Fig. 4B). By inverting the hip joint contact stress onto the femoral head, it was also used to predict the subchondral bone collapse and investigate femoral head reconstruction due to osteonecrosis (Fig. 4C) [21].

Total hip replacement model

A compounded surface and solid model for the hip joint was generated from the Visible Human Dataset to simulate total hip replacement surgery. A proximal femur/hip prosthesis model is incorporated to the pelvic model to study hip range of motion and stress distribution prior to and after hip replacement using different implant designs (Fig. 4D) [22]. The hip implant model was developed using the computer-aided design/computer-aided manufacturing (CAD/CAM) files from the manufacturers or taking the existing implants' plastic replicate for CT scan images. This compounded model allows both cemented and non-cemented hip replacement simulations. Joint range of motion was investigated based on acetabular component

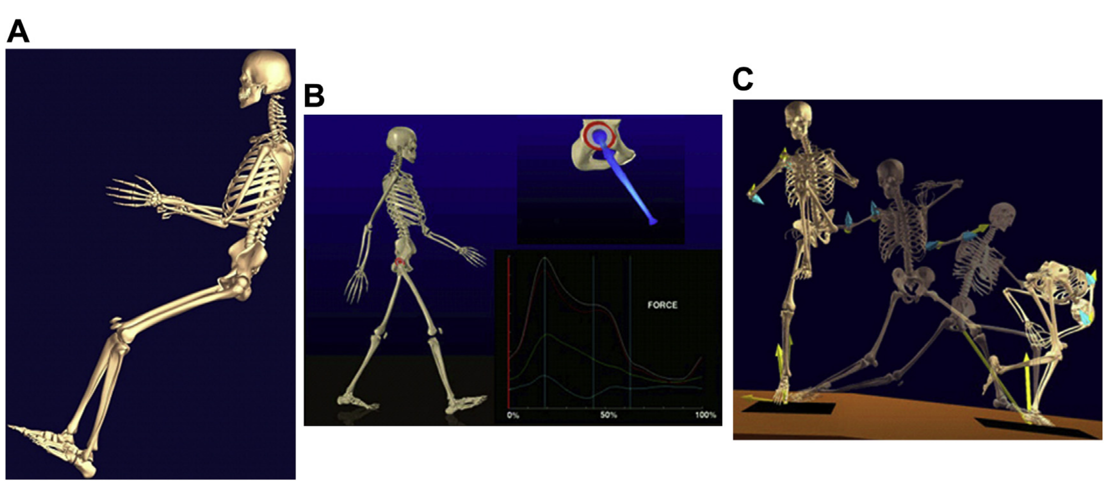

Figure 2 The three-dimensional full-skeleton model of the human used for automobile impact study (A), gait analysis after hip replacement (B) (taken from Lynch JD's MS thesis at Iowa State University, 1994 [12,13]), and the composite view of the full human skeleton to replicate baseball pitching dynamics (C). The shoulder and elbow joint forces (yellow single arrow) and moments (blue double arrow) are shown together with the ground reaction force (yellow arrow) measured by a dynamic force plate for the entire cycle of pitching [29]. An animation segment simulating total hip replacement patient's gait cycle and the associated prosthetic stem stress is available in Video S1 in the Supplementary material online.

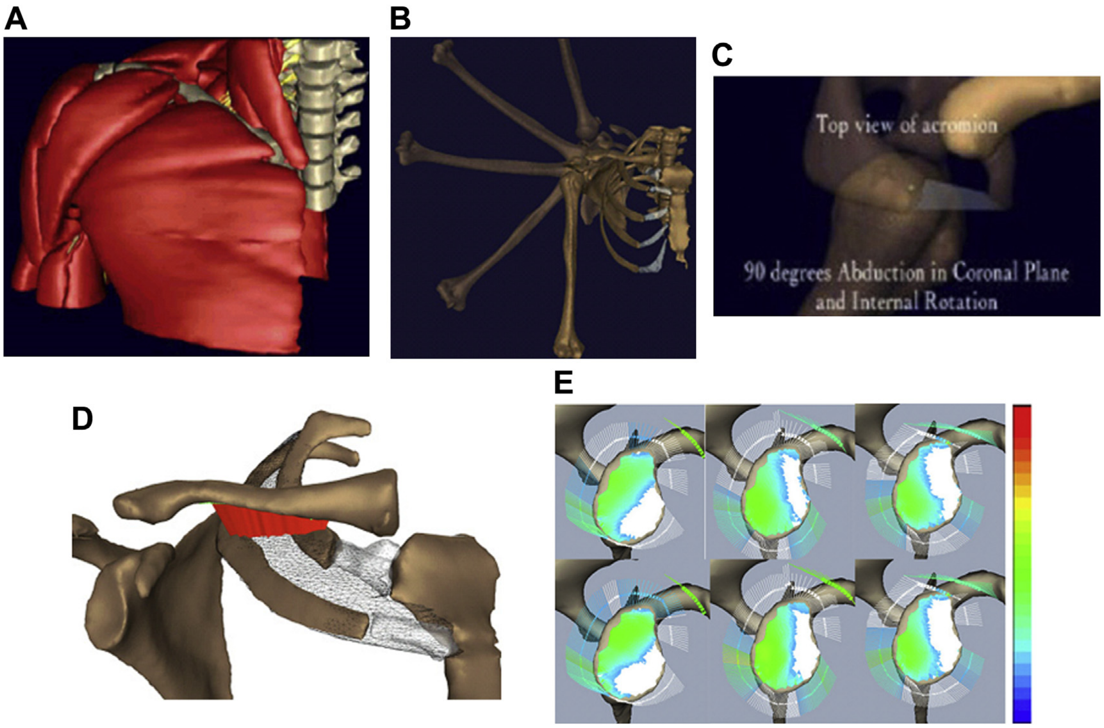

Figure 3 (A) A composite muscular, neurovascular and skeletal model of the shoulder visualized in a sequential manner from the superficial muscles to the underlying bony structure for anatomical studies. An animation segment is available in Video S2 in the Supplementary material online. (B) The sequential images of a cadaver shoulder during passive elevation of the humerus in the plane of the scapula. These shoulder models were created from CT data of cadaver specimens. The kinematic data, measured by using electromagnetic "sensors" (Flock of Birds™, Ascension Technology, Colchester, VT) fixed to the humerus, scapula and clavicle and a "source" mounted on the trunk of the cadaver, was used to animate the shoulder motion rhythm of all the bony structures involved [19]. (C) A solid model of a cadaver shoulder highlighting the history of the closest points between the greater tuberosity and the acromioclavicular ligament during the Hawkins maneuver for impingement test. (D) The same model used to study thoracic outlet syndrome under provocative maneuver tests. The thoracic outlet area between the clavicle and the surface of the 1st and 2nd ribs (marked by the mesh structure) is quantified and highlighted in red color [23]. (E) The glenoid surface model for joint contact area/stress and ligament-capsule tensile stresses study during arm elevation.

placement, joint surface wear, and femoral component neck design. In addition, surgical approach and prosthesis placement were also simulated to illustrate the utility of this model.

Ankle joint contact stress and ligament tension model

3D bone models of the talus, calcaneus, tibia, and fibula based on the Visible Human Dataset (National Library of Medicine) were scaled to match the CT data recorded from cadaver specimens in different joint angles at 10° increments from 30° of dorsiflexion to 50° of plantar flexion covering the entire range of ankle motion during level walking [23]. Regions of potential bony contact were identified by the contour lines of the subchondral bone on each slice of the orthogonal CT sections and were then stacked to create joint contact surfaces (Fig. 5A and B). Rows of tensile strings for the ligaments and the interosseous membrane were inserted at the anatomical regions identified from the dissection data of the same specimen. This marks the first time that the ankle normal contact and ligament stresses have been quantified using biomechanical analysis and simulation (Fig. 5C). This model was used to study a new interpretation of the mechanism of ankle fracture [24].

External fixator and bone fracture reduction, lengthening, and osteotomy model

Two types of unilateral external fixators were modelled as solid rigid bodies of adjustable links interconnected by different joints. Any long bone or pelvis can be incorporated with the fixator forming an open or closed linkage system to study fracture reduction, bone lengthening, and osteotomy adjustment through callus distraction planning using the kinematic chain theory [25,26]. In addition to fixator adjustability studies, this model is now being extended to investigate fixator stiffness performance for device evaluation and design optimization. Finally, an EBI DFS Dimension Fixator was modelled graphically using the CAD/CAM software to demonstrate fracture reduction through fixator joint adjustment for both bridging and nonbridging applications. The parameters of a distal radius deformity were defined from the CT scans and the anterior—posterior and lateral radiographs at the fracture site. Alignment based on the bony landmarks of the radius relative to the intact contralateral side defined the deformity according to dorsal/volar translation, radial shortening, and radial/ulnar translation. Radial and volar/dorsal tilts and axial rotation along the long axis of the radius described the displacement and angulation of the distal

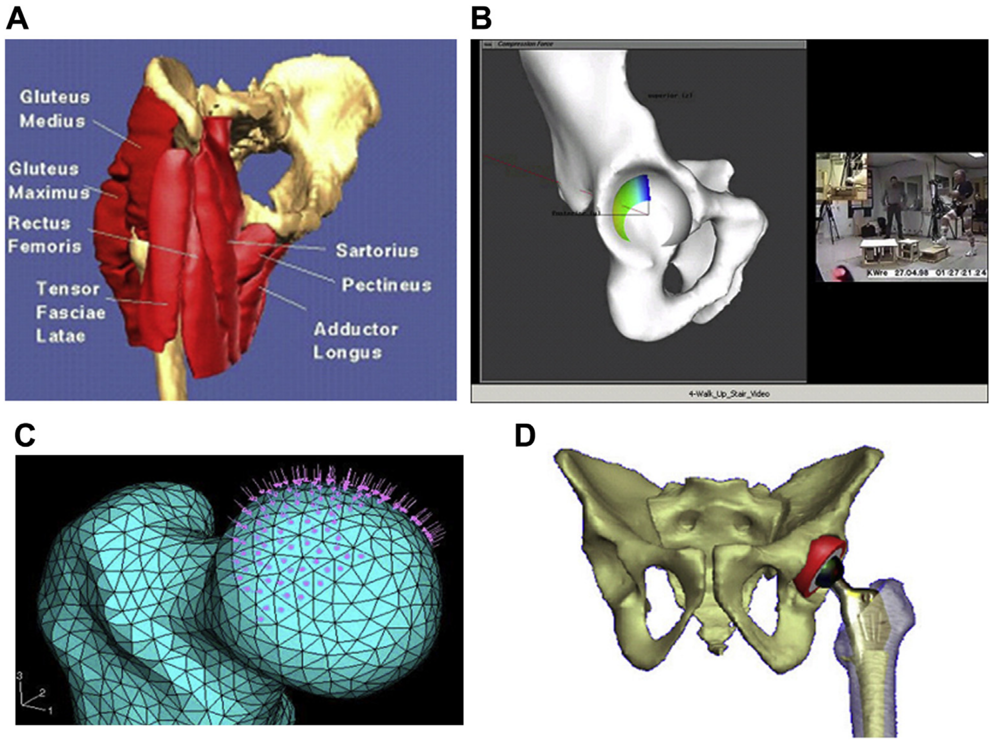

Figure 4 (A) The surface model of the pelvis and the proximal femur with the key muscles across the joint used for the dynamic force analysis of the hip. (B) The model used to study acetabulum contact area and stress distribution during activities of daily living involving the hip [39]. The hip joint reaction force (arrow) and contact stress distribution at three positions during the gait cycle for the left (highlighted) leg calculated using the discrete element analysis (DEA) technique. The blue areas indicate the regions of the lowest stress, whereas the yellow and green regions indicate the locations of higher stresses. (C) The proximal femur model used to investigate subchondral bone collapse due to osteonecrosis (OS) and femoral head reconstruction [44]. (D) The total hip replacement model including the bone and prosthesis components used to study the effects of femoral neck design and implant placement on joint range of motion and potential dislocation. An animation segment illustrating hip range of motion under different femoral neck design is attached, available in Video S3 in the Supplementary material online.

radial fragment. Because the fixator is functioning in a similar manner as a complex robotic arm, the bone-fixator system could be modelled as a multilink closed kinematic chain [27]. A more detailed presentation of this application is given in the section dealing with "VIMS-Lab" to illustrate the use of VIMS in translational research.

Geometric scaling of models

Nearly all models in the VIMS database are generic in nature, and they were developed from the same Visible Human Dataset or the Johns Hopkins Virtual Human database. It would be impractical to use the same laborious process to derive an individual model for a patient for analysis purpose. However, specific bone and joint geometry and dimension can be scaled from the generic model using the acquired X-ray or CT data in order to evaluate the biomechanical effects of the pathology and to simulate the anticipated treatment outcome. Previously, this method was described as the "parametric scaling" technique in the simulation environment using custom software or commercial program such as Pro/ENGINEER (PTC Engineering Solutions, Parametric Technology, Needham, MA, USA). For joint implants and spine and fracture fixation devices, scaling can be accomplished using different CAD/CAM programs. Data for each cross section of the bone can be associated with the plane or its boundary expressed in mathematical forms. Splines used to define the cross section boundary in each plane are modified point by point using the feature-based solid modelling technique.

The VIMS-Model is intended to build a host of musculoskeletal joint generic models that can be manipulated to perform realistic biomechanical analyses on a general population or on individual patients with specific pathologic conditions. When the specific 3D geometry of the patients' anatomy and pathology is required, their CT and magnetic resonance imaging data can be combined with the generic models to approximate the patient-specific model for comparative analysis used to reconstruct the patient-specific model with the added time and cost. Such capability requires further development to make it cost-effective.

When solid models are needed for stress/strain analysis, CAD feature based solid modelling tools are the state of the art. Although the voxel-based models with material property and morphology incorporated are desirable, the surface models [19,28—30] remain as the standards for medical applications. Solid models to fit the finite element method codes for stress analysis can be scaled parametrically to allow the geometry of a bone to be modified to match specific entry data. In this case, the visualization

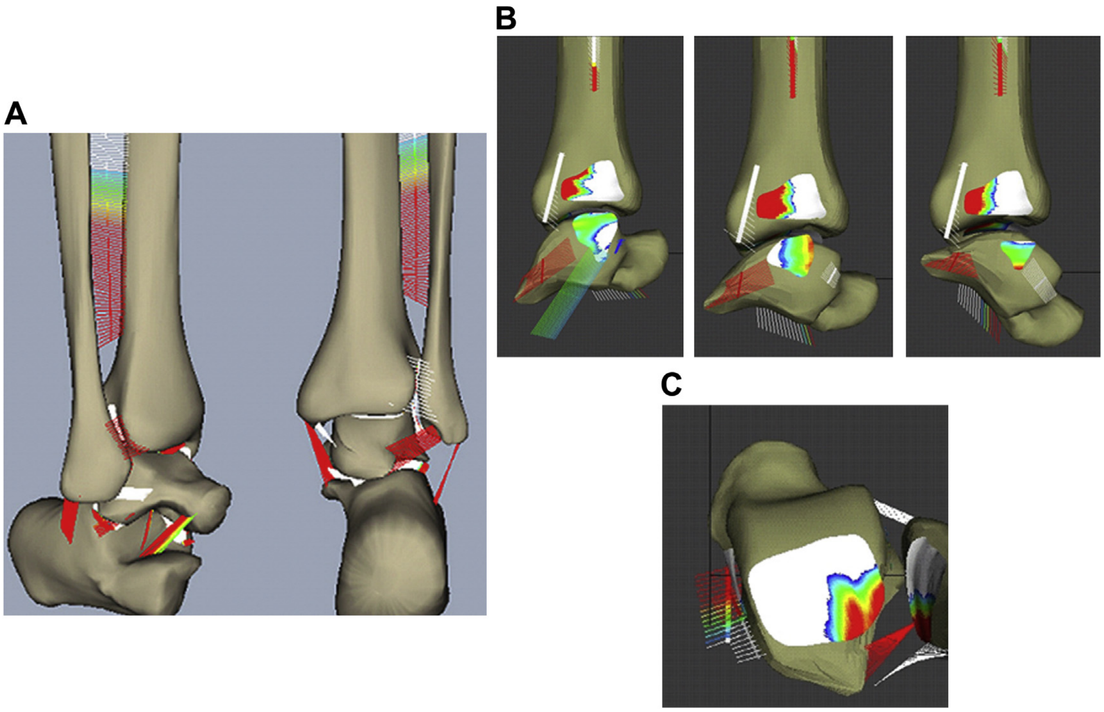

Figure 5 (A) Human ankle joint model of the distal tibia, fibula, talus and calcaneus plus all the surrounding ligaments connecting these bony elements. The ligament tensile stresses are shown as colored lines depicting the their intensity [24]. (B) Ankle contact stress distribution and ligament tension in the tibiotalar and talofibular joints during the stance phase of gait [23]. Left: Heel strike; Middle: Mid-stance; Right: Push-off. (C) Maximum contact pressure distribution at the tibial platfond and ligament tensile stress loaded during the stance phase of gait cycle [24].

of the analysis results will be presented on more refined graphic models.

"VIMS Tool" for biomechanical analyses

Kinematic analysis

In musculoskeletal systems, limb and joint motion is important to define normal functional requirements and the possible pathologic effects caused by joint diseases or neuromuscular abnormalities. Although such information could be observed or measured on living persons, no information could be derived to study the underlying skeletal movement under direct visualization without the effort of attempting to express them in analytical terms using kinematic principles and definitions. Basically, there are two types of motion, the global limb and joint motion and the local articulating surface displacement. The global motion both in translation and rotation can be quantified with fair accuracy using any of the motion analysis systems or externally mounted linkage systems. However, joint articulating surface motion is extremely difficult to measure and visualize. Therefore, the modelling and analysis capability in VIMS will be limited to global joint motion.

Joint rotations in 3D are expressed in terms of the familiar Eulerian angles to facilitate musculoskeletal dynamic analyses and for movement animation. There are two most frequently used systems for Eulerian angle definition, the "three-axes" system and the "two-axes" system. However, the well-known "gyroscopic" system can be used to describe the unique Eulerian angles, which will be rotational sequence independent as applied to the use of external linkage measuring devices for joint motion [3,31,32]. The use of the latter system is usually for the purpose of avoiding the ambiguity of rotational reference when two axes become co-liner, the "Gimbal Lock" phenomenon, under a large range of joint motion such as in the shoulder. In two connecting skeletal segments, their relative motion from one position to another can be determined if their localized coordinate axes are defined in reference to an inertial reference frame. This coordinate system was renamed as the "anatomic" axes for the knee joint [33]. Anatomical joints are rarely pure hinge, universal, or ball-and-socket type as commonly used in mechanical systems. There is always a certain amount of translation involved in their movements. Furthermore, finite rotation of a limb segment is sequence dependent. However, joint action usually takes the easiest or an aesthetically attractive path, which provides the least effort while confined by the constraining soft tissues. These well-known engineering principles can now be demonstrated using the VIMS technology. In the study of skeletal kinematics involving finite displacement and rotation, the concept of instantaneous centroid of rotation and the technique of determining it were first suggested by the senior author. This concept was later adapted and extended in the mid-1980s by the late Herman Woltring [34]. Again, this concept can now be validated and visually illustrated using graphic models and animation as a translational research at work!

Bone alignment correction under external fixation can be studied using rigid body kinematic analysis. When bone segments are involved in fracture, osteotomy, or lengthening cases and they are immobilized by an external fixator, the entire system can be modelled as a spatial linkage chain and studied using the movability analysis using the homogenous 4×4 transformation matrix [3]. Such an analysis can aid in device performance evaluation, design modification, pretreatment planning, and adjustment during treatment. The skeletal-fixator system can also be regarded as a structure to study its stability behaviour especially the micromotions that occur at the bone fracture or lengthening site. Hence, external devices are functioning as a robotic device, which can adjust bone ends from the outside while monitoring its stiffness property to ensure the required biomechanical environment for the bone to respond properly. These analysis algorithms are documented in the VIMS-Tool package for specific applications in different anatomic regions. In the section dealing with VIMS-Lab, an example will be presented to illustrate the utility of these analysis algorithms.

Joint reaction forces and moments determination

A technique for quantifying the joint reaction forces and moments has been well developed and widely applied. The algorithms for calculating the reaction forces and moments acting at these joints are based on skeletal models with interconnecting links. The mass, centre of mass, and moment of inertia for the anatomic segments will be estimated or retrieved from the database in VIMS-Model. When only the displacement data of the system are known, the process of calculating joint resultant force and moment was first defined as the "inverse dynamic problem", and the solution could be obtained using the optimization approach [2]. This class of problem occurred later in control engineering and optimization fields, but its definition and solution method originated from the biomechanical problem cited here; thus, it became a well-documented example of reverse translational research in the late 1960s. Another approach is to complete the kinematic dataset through numerical differentiation [35]. Velocity and acceleration of each link will be numerically derived from the measured displacement. If the joint driving moments and the external loading are given, the process of determining the system's kinematic performance is defined as the "forward dynamic problem", in contrast to the inverse type. These two classes of problems are determinate, as the number of unknowns in the system and the governing equations are equal. The VIMS-Tool package can handle these problems at the users' preference. However, if only the system's performance is prescribed, the process of calculating the driving moment at the joints under known external loading is defined as the "synthesis problem", and the solution process involved is quite complicated and requires an enormous amount of computational time even if one is using a supercomputer. At the moment, the VIMS-Tool does not contain the various algorithms required for this class of problem.

Distribution of muscle forces and joint constraints

The muscles acting about a joint will be modelled as force vectors applied along the muscle centroidal lines throughout the kinematic motion history. In VIMS-Model, the key muscles and their properties related to each joint function are documented to facilitate the dynamic analysis formulation. These muscle forces are required to balance the external forces and inertial forces acting about each joint, and they could be estimated from the known resultant moment and constraint forces as described in the previous section.

However, quantifying the individual muscle forces is an indeterminate problem, because the muscles around the joint are redundant both in the agonistic and antagonistic sense; therefore, there are more unknowns in the equations than the number of equations needed to solve them. Optimization techniques will be needed to solve this class of problem with assumed cost function or optimization criteria and constraint conditions. The underlying assumption behind the optimization method is that the central nervous system controls muscle action by minimizing some performance criteria or to satisfy the preset cost function [6,7,36,37]. The system of equations will also be subjected to the constraints that the muscle stresses—expressed as the muscle force divided by the physiological cross-sectional area—are nonnegative and bonded. Several optimization criteria are incorporated in the VIMS-Tool software, and they can be refined and modified according to more up-to-date development or based on investigators' own choices. This is very much an ongoing field, which may benefit from many investigators if VIMS-Net can be used as a chat environment to share ideas for both the basic and translational research.

Intra-articular contact stress and ligament tension

The joint constraint force can be further decomposed into joint contact stresses and ligament tension using the discrete element analysis (DEA) technique [38]. This technique can be modified to accommodate the mismatch in joint geometric shape and to incorporate additional soft tissues such as menisci, labrums, rotator cuff, and the joint capsule. In the analysis, bones are treated as rigid bodies, whereas the articular cartilage and the ligaments are modelled as matrices of compressive or tensile springs [39,40]. Furthermore, to satisfy the theoretical requirements of such an analysis, the system must be kept in static or quasi-static equilibrium and thus allowing only an infinitesimal (or virtual) displacement in translation. The DEA method requires less computational time than finite element (FE) analysis techniques, and it has been shown to provide equivalent results in estimating joint contact or implant/bone interface stresses [41].

Joint contact area will be determined between the two bone surfaces at each functional position. This contact area will be assumed to be midway between the two bony surfaces separated by the cartilage of certain predefined thickness. A compressive spring is placed on the centroid of each polygon on the concave side of the joint oriented normal to the polygon surface. Any spring in its uncompressed length (the thickness of the cartilage) that does not intercept the opposing bony surface of the joint will be eliminated from the contact area. Therefore, the joint contact area represents a subset of the joint articulating surfaces between the two bones. Ligament resting length and location are determined from the anatomic database. A series of parallel tensile springs will be used to model the ligaments or joint capsule to predict their tensile stresses in each joint position. If the ligament contains different bundles, different sets of tensile springs with varying dimensions and biomechanical properties can be modelled. In this complex model, the system of equilibrium equations in matrix form will be formulated using the principle of minimum potential energy on spring deformation under applied external loads and by applying Castigliano's theorem [42]. The resulting indeterminate problem can be solved using a Gauss—Jordan elimination process. The entire computational algorithm will be iterative in nature because each step of joint loading occurs under the equilibrium condition. Whenever the joint compressive springs carry tensile loads or the tensile springs carry compressive loads, they must be removed from the system and the equilibrium analysis must be repeated. An appropriate convergence criterion will be adapted using the least-squares minimization principle for the iterative process. The basic theory of the DEA (originally defined as the Rigid Body Spring System) was developed by Tadahiko Kawai, Emeritus Professor of Tokyo University [43]. Recently, this technique was extended to deal with joint dynamic problems incorporating nonlinear springs to simulate cartilage response to load under steady-state conditions [43].

Bone and implant stress analysis

Using established 3D FE models acquired or developed, the stress/strain and deformation in bone, ligament, implant, and their interfaces can be determined using any commercial FE model software, such as ABAQUS (Hibbit, Karlsson & Sorensen, Inc., Pawtucket, RI, USA) or PATRAN (MacNeal-Schwiendler Corp., Los Angeles, CA, USA), whose FE code can be imported to the VIMS-Model platform to create specific FE mesh using existing CT data. The size and shape of prosthesis models can be changed using the Pro/ENGINEER software (PTC Engineering Solutions) to fit the host bone model. Interface and boundary conditions are handled by using the special element types available in the commercial codes or to be developed and incorporated to VIMS-Tool for special applications. Various postprocessing software can be imported and combined with the model for stress analysis and graphic presentation of the results.

VIMS-Lab—computer-assisted fracture reduction at the distal radius

To best demonstrate the function of VIMS-Lab while seeing how the entire VIMS technology including VIMS-Model and VIMS-Tool was being utilized in its totality and to serve the purpose of stressing the importance of translational research in orthopaedic biomechanics, the MS thesis (Bioengineering Department of the Johns Hopkins University) of the junior author, Jonathan Lim, who passed away in 2005, came into play. This thesis, entitled "Robotic assisted fracture reduction at the radius," has never been published except in abstract form [27] and is presented here in excert form to commemorate Jonathan's great contribution to the field of orthopaedic biomechanics with some of the most attractive graphics developed by him using the VIMS software and database to their full capacity. The contents of this investigation are outlined and summarized (Fig. 6).

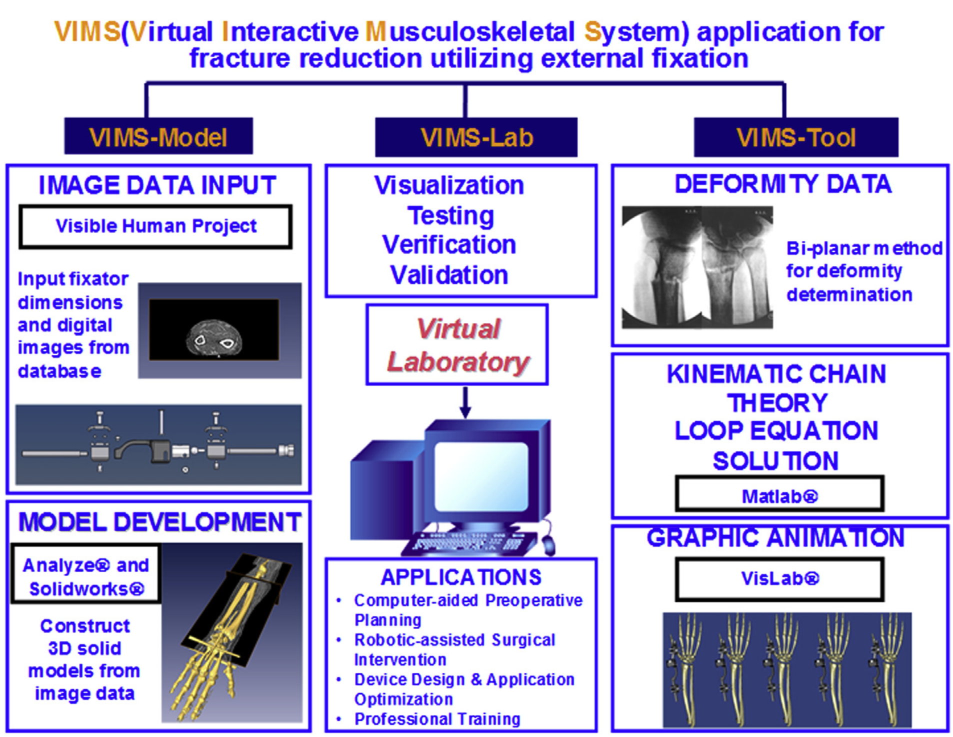

Figure 6 VIMS application outline on distal radial fracture reduction using EBI (a subsidiary of Biomet) DSF™ unilateral external fixator.

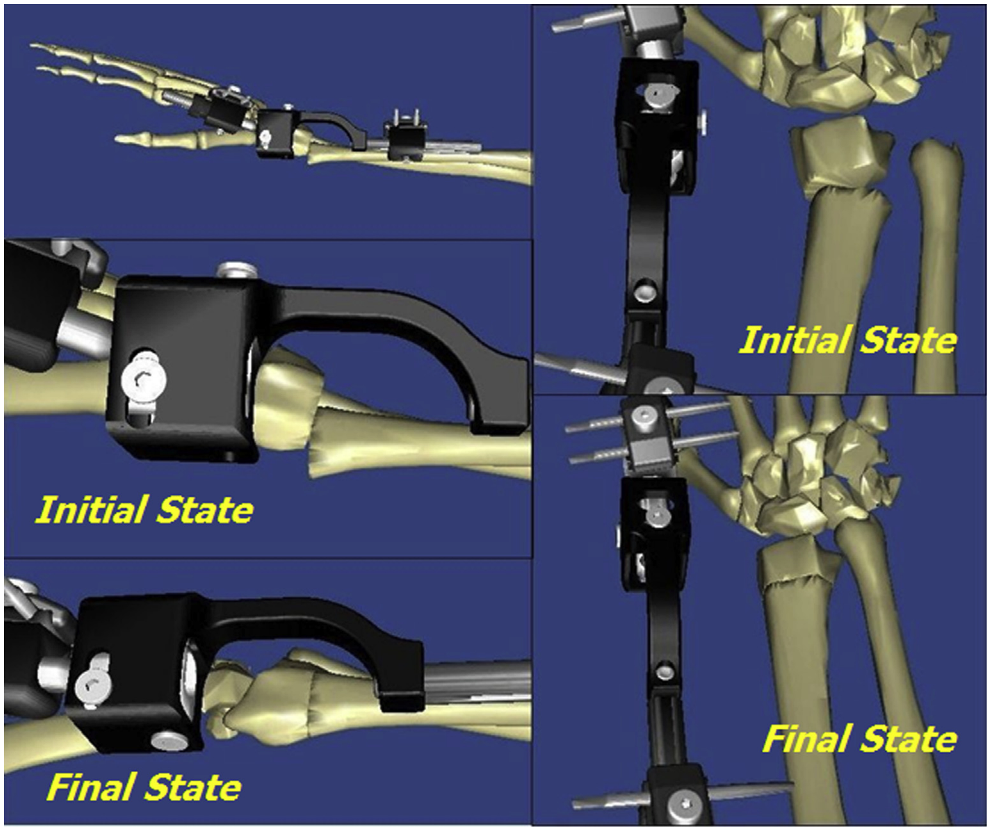

Figure 7 The initial and final fracture and fixator configurations graphic appearance with their close loop linkage pair parameter values shown in Table 1. An animation segment on the simulated wrist fracture event and the sequential in small steps or the simultaneous adjustments (the optimal path) is attached, see Video S4 in the Supplementary material online.

Fracture treatment aims to reduce the bony fragments and maintain that reduction throughout the treatment and rehabilitation period in order to restore functional anatomy. This can be achieved using either internal or external means. However, achieving reduction and fixation externally allows adjustments throughout the treatment course, and when the bone fracture is healed, the device can be removed under minimal intervention. Through simulation, the adjustments necessary to obtain the anatomical alignment can be found. Because the large displacement required is path-dependent, an ideal adjustment sequence to the external device design, very much similar to a robotic device with the least disruption to the underlying vital part, can be found by using VIMS. We therefore hypothesized that a universal external fixator system can be utilized through the optimal reduction path by adjusting the joints of the fixator, thereby achieving the desirable reduction and compression or distraction if necessary without causing any disturbance to the surrounding soft tissues, vessels, and nerves.

An external fixator together with the distal and the proximal forearm bone fragments were considered as a closed-loop kinematic linkage system. It allowed quantification of the translations and rotations required at the connecting joints of this mechanism to reduce a given fracture, provide the desirable gap condition, correct a deformity, and/or lengthen the fracture callus. Combining the kinematic analysis with a graphic model of the bone segments and external fixator would also allow visualization of the adjustments required to reduce a fracture to verify the intended treatment strategy. In the present case, the use of an external fixator coupled with the distal forearm and the wrist through a kinematic chain analysis to prescribe fixator adjustment introduced treatment options that would allow for ideal reduction of a distal radius fracture depending on other clinical considerations. A successful restoration of the normal anatomy of the distal radius is associated with the adequate realignment of the fracture ends with minimal disturbance of the surrounding soft tissue (Fig. 7).

The graphic model for the fixator was developed using Solidworks software. The wrist model was developed based on the CT scan data of the Virtual Human project (Visible Human, National Library of Medicine). From the slice data, a 3D volume was reconstructed by segmentation of the cross-sectional images of the upper extremity using Analyse software (Mayo Clinic, Rochester, MN, USA). A transverse fracture was simulated in the wrist model on the distal portion of the radius. A right-handed coordinate axis was established on the proximal fragment of the fracture with the z-axis oriented along the long axis of the radius and the y-axis oriented in the direction of the palm. The virtual fixator model was applied to the wrist model according to the standard orthopaedic practice in fracture management. One distal pin was secured at the base of the second metacarpal at the metaphysodiaphyseal junction, whereas

$${}^{1}T_{10} = {}^{1}T_{2} \cdot {}^{2}T_{3} \cdot {}^{3}T_{4} \cdot {}^{4}T_{5} \cdot {}^{5}T_{6} \cdot {}^{6}T_{7} \cdot {}^{7}T_{8} \cdot {}^{8}T_{9} \cdot {}^{9}T_{10}$$

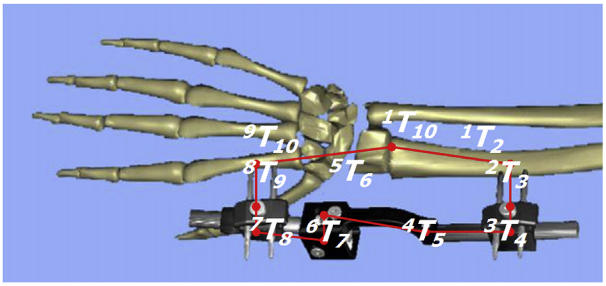

Figure 8 The kinematic chain loop equation as applied to the 10-segment linkage system connecting the hand, forearm and the fixator. The relative orientation of one segment to another as shown by the graphic model and the solid red lines is described by 10 transformation matrices (T) with the connecting links expressed in the left-upper and right-lower subscribes. This matrix equation was solved using the least-square approach using a pre-described convergence criterion.

the other was affixed in the diaphysis region. The proximal pins were affixed 5 inches from the radial styloid at an angle of approximately 45° from the mediolateral plane. The bone-fixator system was modelled as a kinematic linkage chain system allowing rotation and translation at the fracture site through fixator joint adjustments. Based on the final orientation of the bony segments at the simulated fracture site, the solutions could be determined using the coordinate transformation matrix (T) from one link to another, forming the closed loop. An iterative method of least-squares optimization is used to obtain a solution for the joint configurations of the applied fixator to reduce the fracture (Fig. 8). A small distraction was first applied; only three types of joint adjustment patterns were considered although there could be infinite paths that will reduce the finite deformation at the fracture site. The "sequential adjustment" corrects the joints according to the calculated solution values for each fixator joint in a sequential manner. "Sequential incremental adjustment" corrects the joints in sequence in a small incremental fashion. "Simultaneous adjustment" describes a synchronized adjustment of all joints at the same time.

The fracture radiograph and the bone orientation of a clinical case was chosen to demonstrate the reduction to be conducted using VIMS (Fig. 6). The corresponding fixator and fracture site configurations prior to and after reduction are given in Table 1. The number of iterations to arrive at the initial parameters was related to the magnitude of the individual fracture parameters. For the clinically relevant case, the optimization process went through an average of 45 iterations to arrive within 1×10^{-7} of the objective function for the final solution parameters. The convergence time averaged 5 s on a modern personal computer (PC) using the neutral fixator configuration as the estimate for the initial fixator configuration. Among the 10 adjustment parameters (4 translation and 6 rotation), there were only eight required alterations under the given fracture misalignment. These adjustment solutions can be implemented in any combination, sequence, and magnitude, which will lead to the same final reduction. However, each of these infinite combinations will lead to a different correction path, because finite displacements are path-dependent.

The maximum deviation in the axial plane at the fracture site was recorded for different adjustment paths because it represents the potential soft tissue disruption that occurred during the manipulation. The sequentially adjusted case (each fixator adjustment parameter was implemented to its magnitude in one step) resulted in a maximum deviation of 45.7 mm in the system, which would render it inadmissible due to the unreasonable degree of translation at the fracture. When the individual adjustment was halved, the maximum deviation was 31.1 mm, which was still unacceptable. Finally, if all adjustment parameters were simultaneously adjusted in equal steps of greater than 10 or each parameter was implemented in small magnitudes of less than 1 mm or 1°, the maximum deviation was reduced to 5.1 mm. In fact, the last two adjustment methods had created nearly identical paths, and that would converge to one single unique solution with increasing step or decreasing stepwise adjustment magnitude. This unique path is defined as the optimal path achievable by rotating with respect to the instantaneous screw axis.

For any given fracture characteristics and pin placement in a reduction procedure, there exists an optimal adjustment strategy for the fixator joints among all correction paths. In a unilateral fixator design like the one discussed here, rotational deformity may not achieve an ideal reduction. Therefore, an appropriate manual reduction in

Table 1 Fixator and fracture site configurations prior to and after reduction using VIMS.

Initial state				Final state			
Fracture site		Fixator joint configuration		Fracture site		Fixator joint configuration	
d_x	0 mm	X_1	5.02°	d_x	0 mm	X_1	5.02°
d_y	4 mm	X_2	15.7 mm	d_y	0 mm	X_2	16.2 mm
d_z	−1 mm	X_3	5.08°	d_z	0 mm	X_3	−43.2°
R_x	−15°	X_4	−0.07 mm	R_x	0°	X_4	2.26 mm
R_y	−10°	X_5	4.73 mm	R_y	0°	X_5	7.63 mm
R_z	2°	X_6	13.05°	R_z	0°	X_6	16.78°
		X_7	0.86°			X_7	−5.3°
		X_8	8.27°			X_8	51.52°
		X_9	1.79 mm			X_9	−10.47 mm
		X_{10}	13.92°			X_{10}	13.92°

axial rotation is recommended prior to mounting the fixator for the final reduction and manipulation. This minor limitation should not affect its use in osteotomy or lengthening applications. An alternative fixator, the Taylor Spatial Frame (TSFR by Smith Nephew, Memphis, TN, USA) [44], uses six linear adjustment cylinders to provide larger rotational deformity corrections similar to a flight simulator design. The DFS fixator contains one ball-and-socket joint with 3 DOF (degree of freedom) not easily controlled for individual in adjustment in each axis of rotation. However, single stepwise rotations with respect to the instantaneous screw axis can be instrumented to provide smooth corrections with minimum deviation of the bony segment very similar to human natural joint functions. This could be adapted in the robotic field to provide a bionically featured joint that will reduce the bulk of the machine and provide sooth motion to save power while achieving the aesthetic movement of a robot in the game or entertainment industry. This serves as the final example in reserve translational research related to biomechanics, that is, using human system structure and function to enrich engineering.

Analysis validation

An adequate validation of the virtual models used in VIMS technology is essential to establish its credibility. The Dynamic Knee Simulator (Fig. 9), developed by the author in closed collaboration with MTS System Corporation (Eden Prairie, MN, USA), was used to validate the model and analysis algorithms related to the knee joint [45]. On this simulator, the anatomical specimens were instrumented to measure a simulated squatting activity under gravitational load of the body. The measured joint contact pressure and the bone internal force and moment were compared to those calculated using the VIMS-Tool computational algorithms and the generic knee model available in the VIMS-Model library after an appropriate scaling.

Another way to verify the model and analysis results is to examine the graphic output results in the VIMS-Lab environment. As in the kinematic, kinetic, and the anatomical studies using the virtual but realistic and geometrically accurate models, one can visualize and perceive with reasonable confidence whether the final results of the analyses are correct and reliable [16,26,46]. However, the aesthetic appearance of the virtual models and an absolute validation of the analysis results on internal force, stress/pressure of the musculoskeletal system in action would be difficult but unnecessary as long as the trend of the results matches that measured on the simulator under a similar loading regime or we can compare them qualitatively according to published data and trained judgement. An additional test setup involving other joint models and analysis conditions could be developed to provide an overall qualitative assessment of VIMS-Model, VIMS-Tool, and VIMS-Lab, but that will be time-consuming and will not really add much to the fundamental value of simulation in the field of biomechanics. Moreover, the final proof of the validity of the results from VIMS simulation will rely on appropriately designed clinical trial studies similar to any biological or pharmaceutical translational research outcome.

Discussion

The "Virtual Human" is an exciting reality for biomechanical analyses and simulation well demonstrated in this paper using the musculoskeletal system as an example. With further development, this technology shall become a broad foundation with full-featured analysis capability, robust model library and database, and a well-organized laboratory environment to serve as a biomechanical simulator for a wide spectrum of basic science and clinical applications. This simulation technology unites the expertise in biomechanical analysis and graphic modelling to investigate joint and connective tissue mechanics and to visualize the results in both static and animated dynamic forms together with the system involved. Adaptable anatomical models including implants and fracture fixation devices and a computational infrastructure for static, kinematic, inverse and forward dynamic, joint contact pressure, stress and strain analyses under varying or moving boundary and loading conditions that are incorporated on a common software platform comprise a novel, timeless, and significant advancement in the field of musculoskeletal

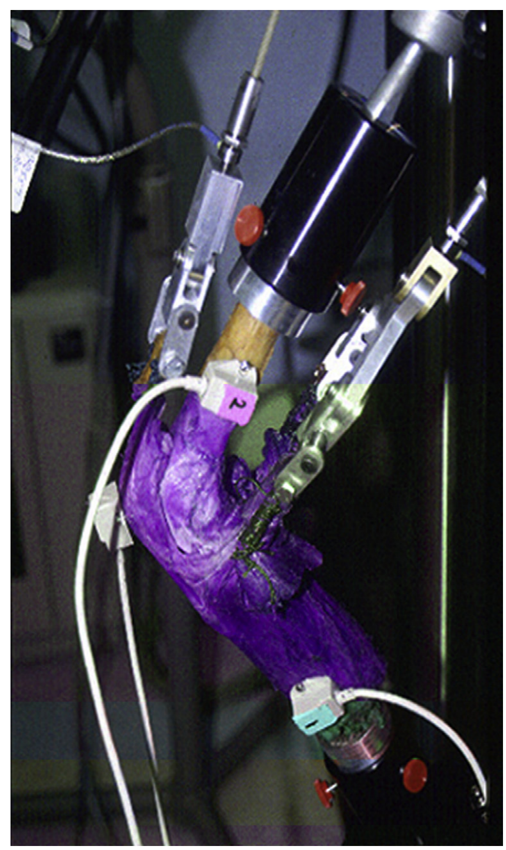

Figure 9 The Dynamic Knee Simulator used to study knee flexion and joint loading under simulated squatting activity. Independent loads are applied to the simulated hip joint, the medial and lateral hamstrings tendons and the quadriceps tendon using hydraulic actuators. The tendons are secured to the loading actuators using cryoclamps. The MTS Model 790.00 TestStar II Control System software (MTS Systems Corporation, Eden Prairie, MN, USA) was used to control and monitor all motion and loading conditions [30].

biomechanics to provide the needed impetus to revive interest and emphasis in this particular sector. Although there are several model-making software packages available on the commercial market, nothing comes close to VIMS's capabilities, and some of the intended applications are demonstrated in this report. These commercial products often exaggerate their clinical utility in performing treatment plans, but without biomechanical analysis and justification, these planning efforts cannot be established on firm ground. In relation to translational research, VIMS should be regarded as a practical technology to put biomechanics in the service arena similar to anatomy and physiology in rendering safe and effective treatments.

In engineering, translational research has always been the fundamental requirement as opposed to the basic science research in biology, chemistry, physics, and mathematics. Although conventional clinical trial studies are not required, testing programs under unforeseeable natural disastrous conditions and man-made errors are included to strengthen the design and manufacturing requirements. In addition, generous safety factors are usually included as built-in features of more critical machines such as aircraft, nuclear power plants, and high-speed vehicular equipment to ensure the safety of workers and passengers and to secure safe and desirable working conditions. In medically related technology, because of the unpredictable environment that the patients are facing, the safety criteria are far less desirable. Hence, meticulous clinical trial studies are mandatory after translational research. Bioengineering procedures and products will face the same regulations.

This simulation technology will in no way completely replace the need to conduct experimental testing using human and animal anatomical specimens mounted on universal testing machines or custom-made joint simulators. Although time-related simulation on material fatigue failure or tissue growth and remodelling has been performed in the past, animal studies are still regarded as the standard method in bone and joint research and implant development. The results generated from all of these studies, experimental or theoretical, shall be validated by randomized controlled clinical trials to prove their safety and efficacy. What the VIMS can offer is a set of generic models and parametric analysis for comparative purposes in healthy and patient populations to strengthen the results of controlled clinical trials and stratify further studies if indicated. In individual patients, it also provides the unprecedented capability of assisting physicians and surgeons to optimize treatment protocols to improve clinical outcomes and minimize risks.

The simulation software and database in VIMS were developed for the purpose of enhancing research, education, and clinical patient care related to musculoskeletal joint function at the structure, organ, and system levels. Little effort has been devoted to model and analyse connective tissue at the material level except for the cartilage on joint contact problems [21,43]. However, this can be included whenever these tissues' nonlinear and time-related behaviours are well formulated. Likewise, long-term growth and remodelling of normal, repaired, or reconstructed systems can be simulated as demonstrated in the past [47,48]. Therefore, VIMS at its current development is limited to structural analyses of the musculoskeletal system to provide the front-end data, which could be used later for the downstream tissue level modelling and analysis purpose. It would be desirable, however, if the analysis tools for muscle force determination could include some neuromuscular control theory so that future simulations of the musculoskeletal system can be extended to include synthesis problems related to its physiological performance. From the clinical point of view, this technology should have strong appeal to both patient care and rehabilitation training with its unique graphic-based models and computer animation of the biomechanical responses to loading and motion under normal and pathological conditions. The limitation on model storage and lengthy computational time required can now be handled in the CC environment, which will be discussed in the following paragraph. Hence, VIMS can be made available to users (using a PC and its operating system and graphic software) at an affordable cost.

Several computational algorithms and model library databases were integrated into the VIMS system platform on SGi supercomputer mainframe using the Unix operating system. All of the independent analysis components of the software are accessible through a single graphical user interface. This software package can be modified to fit the X-Windows/OS PC operating system. Users in the public domain can access the VIMS-Model and search through the model library to select the desirable musculoskeletal region and the orthopaedic implant or device for the intended simulation and analysis. The kinematic data of the anatomic system involved in functions of daily living or sports activities could be adapted from the literature or measured to serve as the input data for biomechanical analysis on the generic models. The analysis results will be graphically presented and animated using the VisLab software (Engineering Animation Inc.). Unfortunately, this utility software plus the VisModel package are no longer being served by the commercial firm, and they need to be converted to a PC-based operating system in order for the VIMS to gain acceptance and popularity. To facilitate easier access and utilization of VIMS, it is necessary for an organization (either private or institutional) to streamline the maintenance, operational procedures, and protection of the system on different PC operating systems on a service fee-based arrangement. Again, to alleviate the limitations on core storage and computational speed in PCs, the CC environment would make VIMS more accessible to private and industrial/institutional users.

This integrated system will no doubt make the learning of functional anatomy easier and help create virtual laboratories on the Internet to share the resources, analysis algorithms, and research findings. Such capabilities will expand the scope and utility of musculoskeletal biomechanics without relying on the use of animals or cadaver specimens despite the restrictions posed by the limitations of models and loading complexity. This broad-based technology will not only revolutionize the development and testing of orthopaedic implants and devices to improve their clinical performance and reliability, it will also make biomechanics competitive in landing federal funding and industrial contracts. Finally, the development of biomechanically justified preoperative planning strategy and the associated execution procedures and operational steps under a VR environment using accurate and realistic graphic models combined with

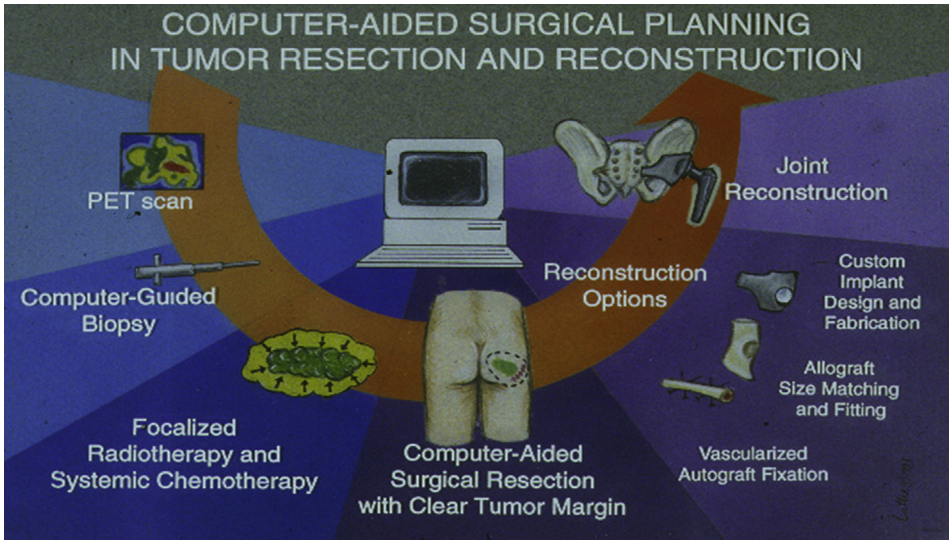

Figure 10 The concept proposed for computer-aided surgical planning in musculoskeletal tumor resection and reconstruction for limb salvage in the late 1980s before the emergence of the VIMS technology. In this diagram, the current technologies on navigation and robotic-assisted surgery were incorporated. In addition, accurate imaging and biopsy for tumor diagnosis, focalized radiotherapy and other local tumor ablation techniques such as microwave radio frequency ablation, cryosurgery, etc. and systematic chemotherapy were included. Today, with the VIMS technology available for individual patient's lesion model making, therapy and surgery simulation, safer and more effective limb salvage procedures can be delivered.

biomechanical rationales will provide the essential foundation and tools for the true computer-aided surgery currently active in the fields of surgical navigation and robot-assisted surgery. An important spin-off utility of these emerging technologies is in the field of limb salvage in orthopaedic oncology (Fig. 10). The application of radio frequency ablation or hyperthermia in the treatment of bone tumour would be an ideal application of VIMS both in treatment planning and execution to make limb-saving surgeries safe and effective and provide improved and durable limb function. Other possibilities of adapting VIMS to other medical applications such as computer-aided rehabilitation are only a few steps away from reality. To conclude, simulation technology, although an old and powerful engineering discipline, is able to find its way in orthopaedics translational research with both proven and unexplored utilities. Under the CC and virtual storage environment, it will definitely make biomechanics laudable and easier to apply, and thus appreciated by clinicians. Finally, through the experiences and advances gained in the more challenging biomedical fields, a reverse translational research will be achieved to not only elevate the status of biomechanics but also enrich the engineering field.

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The development of the present VIMS technology involved many staff and fellows both at the Mayo Clinic and the Johns Hopkins University, too numerous to properly acknowledge here. However, Jonathan Lim was the exception as his untimely death prevented him from publishing his MS thesis work and becoming a physician, as he would have wanted. We all miss him enormously, especially for an aging teacher still lamenting the tremendous loss of a dear student and friend! The initiation of this developmental program was made possible by a subcontract from EAI (Engineering Animation Inc., Ames, IA, USA) through an ATP (Advanced Technology Program) grant awarded by NIST from 1993 to 1996. In the ensuing years, the VIMS software refinement and its application expansion were partially supported by the Orthopaedic Research and Education Foundation through its Bristol-Myers "Center of Excellence Grant," by a major private donation from the Nobuhara Hospital in Tatsuno, Japan, and by generous gifts provided by the Industrial Technology Research Institute of Taiwan, the EBI Medical Systems (now a subsidiary of Biomet) and a private industrialist, devoted benefactor in charity work worldwide, and close friend, Dr. In-Long Chu from Taiwan.

Appendix A. Supplementary data

Supplementary data related to this article can be found online at http://dx.doi.org/10.1016/j.jot.2013.07.006.

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